NASA TECHNICAL NOTE

NASA TN D-6601

NASA TN D-6601

DIGITAL AUTOMATIC DATA REDUCTION TECHNIQUES USED IN A 1000-FLIGHT BIOMEDICAL STUDY

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NATIONAL AERONAUTICS AND SPACE ADMINISTRATION • WASHINGTON, D. C. • DECEMBER 1971

1. Report No. NASA TN D-6601	2. Government Accession No.	3. Recipient's Catalog No.
4. Title and Subtitle DIGITAL AUTOMATIC DATA REDUCTION TECHNIQUES USED IN A 1000-FLIGHT BIOMEDICAL STUDY		5. Report Date December 1971
		6. Performing Organization Code
7. Author(s) Richard Carpenter and James Roman		8. Performing Organization Report No. H-651
9. Performing Organization Name and Address NASA Flight Research Center P. O. Box 273 Edwards, California 93523		10. Work Unit No. 127-51-03-01-24
		11. Contract or Grant No.
12. Sponsoring Agency Name and Address National Aeronautics and Space Administration Washington, D. C. 20546		13. Type of Report and Period Covered Technical Note
		14. Sponsoring Agency Code
15. Supplementary Notes		

16. Abstract

Techniques developed to automatically process a large quantity of physiological data obtained during a 1000-flight study are described. To reduce this data reliably, a study program was conducted using physiological data from X-15 flights as a data source for experimenting with signal enhancement and noise elimination techniques.

The techniques include an automatic means for counting heart rates, averaging electrocardiogram waveforms, plotting histograms of heart rate versus frequency, and counting respiration rates. These techniques were used to reduce more than 2000 hours of physiological data recorded in flight.

17. Key Words (Suggested by Author(s))	18. Distribution Statement
Physiological data Automatic processing Techniques	Unclassified - Unlimited

19. Security Classif. (of this report)	20. Security Classif. (of this page)	21. No. of Pages	22. Price*
Unclassified	Unclassified	15	$3.00

*For sale by the National Technical Information Service, Springfield, Virginia 22151

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INTRODUCTION

Modern sensing and recording techniques have made it possible to record physiological information continuously during flight (refs. 1 and 2). Continuous physiological information has also been collected over the years during centrifuge programs. The use of continuous monitoring has resulted in the accumulation of vast quantities of data. Because manual data reduction is prohibitively expensive, except for selected portions of data, and because computer programs specifically designed to handle and reduce large-scale collections of continuous physiological information were not available until a short time ago, most of these data have never been completely reduced. In fact, some have yet to be viewed.

The use of computers is the only practical way to obtain information from medical data recorded continuously in flight; thus, the development of pertinent data handling, data reduction, and data evaluation techniques is of importance to aerospace medicine and, perhaps, to clinical medicine.

Techniques were required to process physiological data obtained in a low signal-to-noise-ratio environment. Thus the possibility of manually reducing the data obtained in a 1000-flight study from oscillographic tracings was investigated. After a few simple calculations it became apparent that the amount of data accumulated during the flights, each with a duration of 100 minutes, would be overwhelming: of the order of 1×10^7 heartbeats and approximately 1×10^6 respiration cycles, for example. Therefore a concerted effort was made to provide the investigators with an easily interpretable format for each flight. This paper discusses the computer techniques developed at the Flight Research Center to reduce the physiological data which were collected in over 2000 hours of flight.

SYMBOLS

N	number of ECG cycles averaged
$r(\tau)$	normalized autocorrelation function of the ECG cycle
T	time period, sec
t	time, sec or min

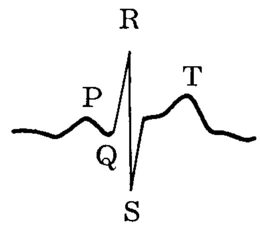
$x(t)$	ECG signal
σ	noise power
τ	time lag between data points, sec
$\psi(\tau)$	autocorrelation function

Subscripts:

av	average
max	maximum
1	first $r(\tau)$ full waveform

Electrocardiographic terms (see adjacent sketch):

P-wave	contraction of the atria
QRS complex	contraction of the ventricles
T-wave	relaxation of the ventricles



GENERAL SYSTEM DESCRIPTION

A literature search did not reveal any techniques or hardware specifications for reducing physiological data obtained in a low signal-to-noise-ratio environment. Therefore, investigation was required to determine applicable data reduction techniques.

Investigations were conducted using a system consisting of an on-line computer which had direct conversation capability between the investigator and the computer, together with the capability for reading in large quantities of data. This system consisted of a high-speed computer at a location remote from the user and terminal equipment (i.e., input and output equipment) at the investigator's location. The input-output data flow was accomplished via telephone lines to the computer complex. This type of on-line computer greatly facilitated the development of the automatic data processing technique.

The on-line computer used QUICKTRAN language, which is a subset of FORTRAN; however, later, when the size of the digital reduction program was fully realized (approximately 4000 cards), the program was rewritten in machine language which reduced the computer running time per flight by a factor of four. At present, the digital reduction program exists in both the IBM 7094 and IBM 360 assembler language and processes a 100-minute flight in 35 minutes on the 7094 computer, and in 170 minutes on a 360-50 computer.

AUTOMATIC DIGITAL REDUCTION OF THE ELECTROCARDIOGRAM

The raw electrocardiogram (ECG) signal obtained from pilots in flight (ref. 1) required some form of enhancement, in that it was anticipated that body artifact and the aircraft environment would often cause a signal-to-noise ratio of 1 or even less than 1 during flight. It was realized that filtering alone would not be sufficient because the frequency of noise encountered would be within the ECG frequency band. Autocorrelation of the ECG waveforms to determine the heart period was chosen as the most promising technique because any periodic data, such as ECG data, have an autocorrelation value for all time displacements, τ, and the autocorrelation function for all noise diminishes to zero for large τ. However, to use this technique it was necessary to detect a reference point for counting. This was possible since one of the characteristics of autocorrelation ($r(\tau)$) of the ECG cycle is that the function has a maximum for each ECG cycle. Hence, the time difference between any two $r(\tau)$ maximums would indicate exactly the period between heartbeats. Locating each maximum with close resolution required two steps: (1) a search mode searched for the maximum $r(\tau)$ values computed from narrow band ECG data in large steps (24 milliseconds); and (2) within the time boundaries established by the first search mode, a fine detection mode established the peak within 3 milliseconds resolution. The fine detection mode used $r(\tau)$ values calculated from wide band ECG data. In effect, the search mode established the time boundaries for the fine resolution autocorrelation. From this point the automatic computations established the time difference between the finely resolved $r(\tau)$ peaks, which provided the period between any 2 cycles. These periods were then averaged over minute intervals to obtain heart rates.

The reliability of the search mode was greatly enhanced by using narrow band (1 Hz to 15 Hz) ECG data which provided wide $r(\tau)$ maximum peaks with high amplitude and good spread. Also, periodic noise spikes were partially eliminated by including logic that prevented $r(\tau)$ spike detection except in the time range of 240 milliseconds to 1440 milliseconds (corresponding to heart rates between 42 and 250 beats per minute) immediately preceding a maximum peak detection. Data reliability was further increased by not including ECG data that autocorrelated at values of less than 0.4; hence, only those ECG cycles that autocorrelated with values of 0.4 or greater were considered in establishing heart rates. Each flight was presented by the computer output format in terms of heart rate per minute, where the average heart period was calculated as the sum of heart periods divided by the number of corresponding cycles that autocorrelated above 0.4. The number of omitted beats was totaled and printed out for each minute interval. Plotted simultaneously with the heart rate for each flight were the maximum, mean, and minimum values of the autocorrelation values during each minute of the flight. This provided a means of estimating the degree of random noise being encountered, which, in turn, provided a rapid estimate of data reliability. A sample of the flight data is shown in figure 1.

Autocorrelation Function

The definition of the autocorrelation function, $\psi(\tau)$, for the ECG time function,

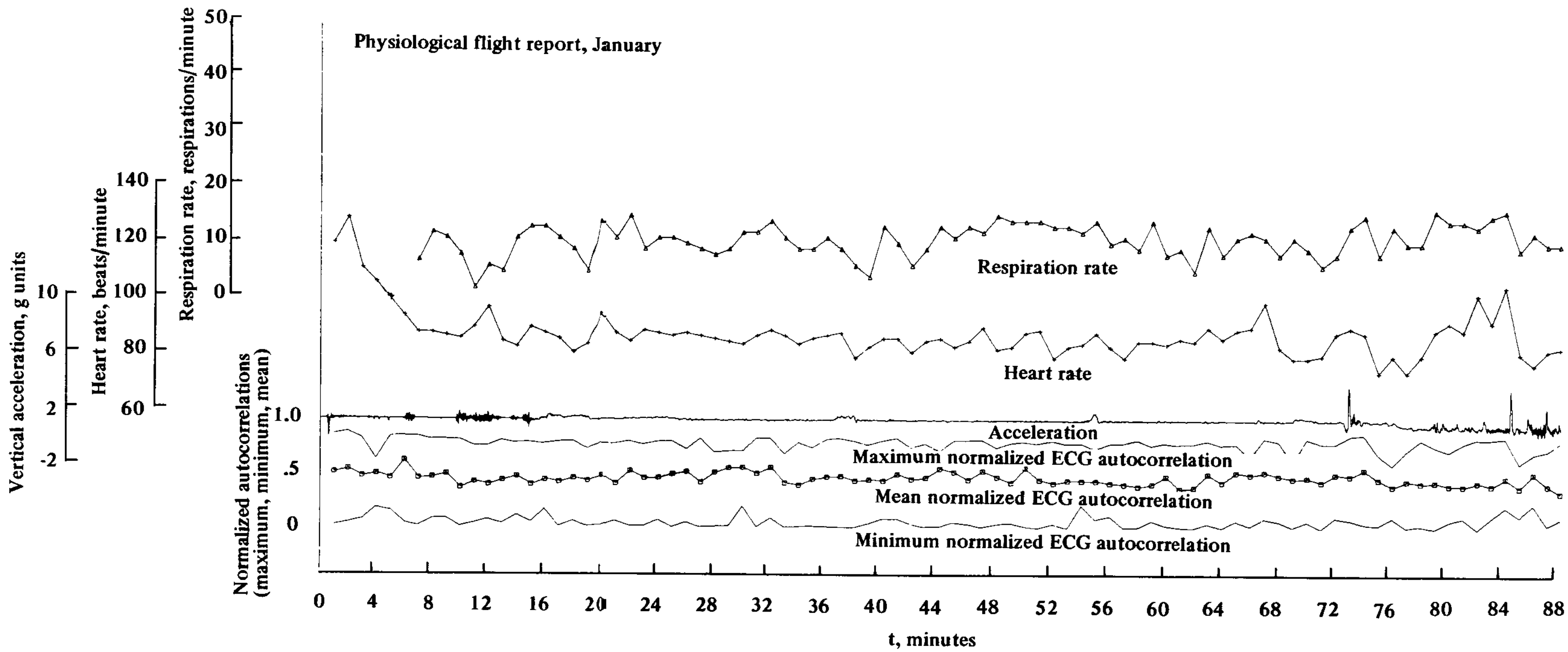


Figure 1. Automatically calculated and plotted physiological flight data.

x(t), is given by

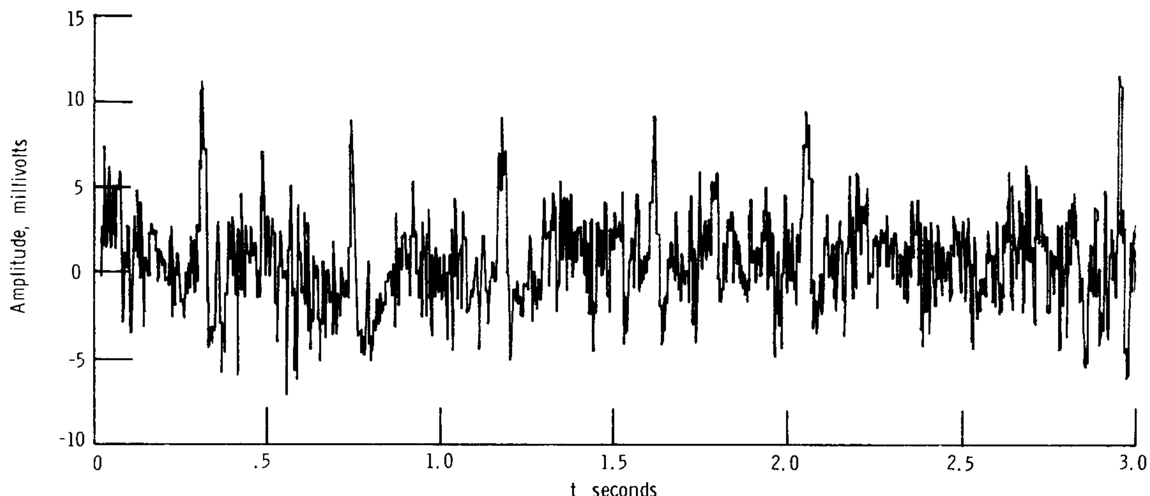
$$\psi(\tau) = \lim_{T \to \infty} 1/T \int_{-T/2}^{T/2} x(t)x(t + \tau)\, dt$$

where T is the period being considered and τ is the time lag between data points. However, to make the analysis and technique independent of gain variations and absolute power levels in the signal conditioning chain, it was found best to introduce and work with the normalized autocorrelation function, $r(\tau)$, which is simply defined and calculated from $\psi(\tau)$ as follows:

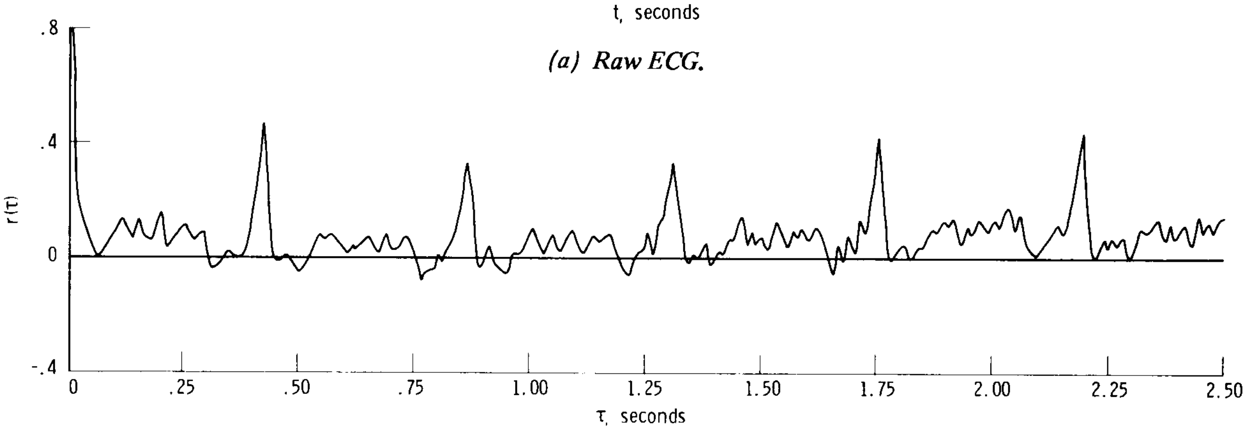
$$r(\tau) = \frac{\psi(\tau)}{\psi(0)}$$

This expression provides a value of unity for the normalized autocorrelation at zero lag.

To evaluate the application of autocorrelation to cardiotachometry, many experimental runs were conducted. An ECG segment from an X-15 flight, on which a Mach number of 5 and a maximum altitude of 60.4 kilometers were obtained, is presented as an example. The ECG plot, which exhibits heavy noise, is shown in figure 2(a). The normalized autocorrelation plot is shown in figure 2(b). Despite the large amount of



(a) Raw ECG.



(b) Normalized autocorrelation of ECG.

Figure 2. Comparison of autocorrelated ECG with raw data.

noise, particularly high frequency, which produces false R-waves in the ECG, the 1-per-heart-cycle peaks of $r(\tau)$ are clearly discernible. This demonstrates the ability of the autocorrelation technique to discriminate signal information reliably even with a signal-to-noise ratio of less than 1. In figure 2(b) the heart rate and signal-to-noise ratio are calculated as follows:

$r(\tau)$ peaks six times at $\tau = 0,\ 0.428,\ 0.875,\ 1.318,\ 1.76$, and 2.21 seconds.
Peak-to-peak lag intervals are 0.428, 0.447, 0.443, 0.442, and 0.45 seconds.

Average period is $\frac{2.210}{5}$, 0.442 second.

$$\text{Heart rate} = \frac{60}{0.442} = 135 \text{ beats per minute.}$$

$$\text{Signal-to-noise ratio} = \frac{r(\tau)_1}{r(0) - r(\tau)_1}.$$

τ = average period, 0.44 second.
From figure 2(b), $r(\tau) = 0.46$ at $\tau_1 = 0.44$ second.

$$\text{Signal-to-noise ratio} = \frac{0.46}{1 - 0.46} = 0.85.$$

The total computation time was considerably decreased by establishing the expected period and reducing the maximum lag, τ_{max}, to be just larger than this value. Computation times were further reduced by integrating over a time increment of just less than 2 periods. This was possible because the decision was made to determine beat-to-beat periods rather than multibeat averages.

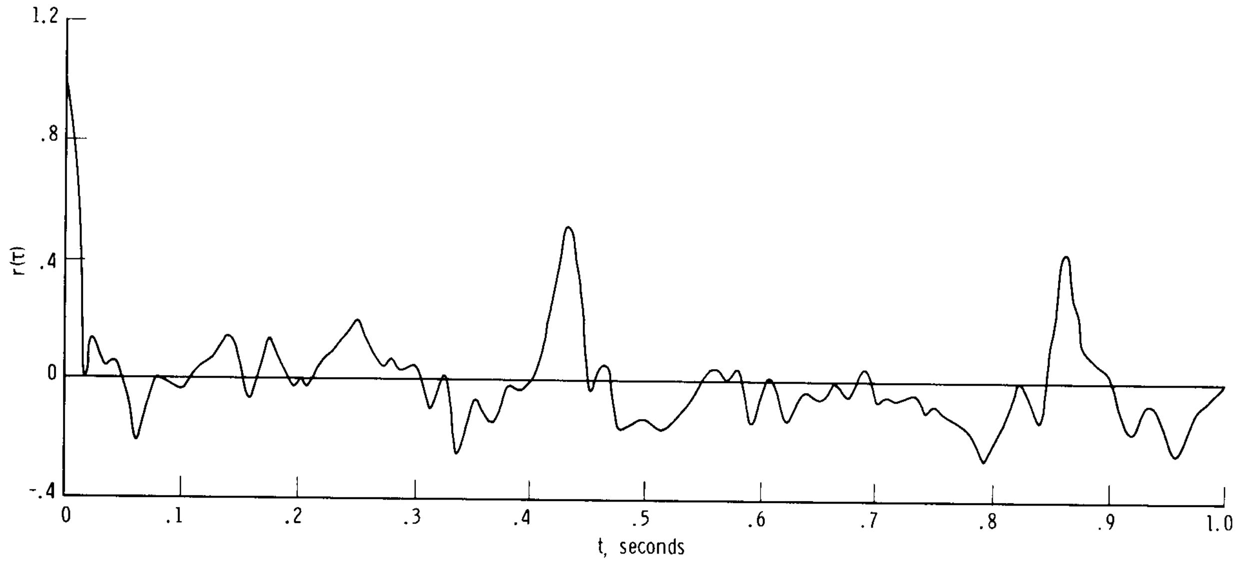
The general operation of the program is as follows: a sample ECG input is passed through a digital bandpass filter and then autocorrelated. The cycle is detected by locating the full peak in $r(\tau)$, which may be a subpeak or the maximum peak, through a coarse resolution search process using filtered ECG data. Following cycle detection, the period of the ECG is determined by locating the $r(\tau)$ peak first at large error with filtered ECG data, then at low error with unfiltered data. The unfiltered and filtered inputs are then updated by advancing the inputs by a number of sampling intervals corresponding to the lag of the peak. Then the next cycle is ready for processing, after which a power change test between peaks is made. If the power change is large, the next cycle (peak) detection is performed with the search mode, but with the next time interval of ECG data as an autocorrelated input. Thus, the program continues. The amplitude of the autocorrelation of the wide band data input at the peak (the time at which the cycle occurs) of the cycle and the lag of the peak are produced as the output. However, if the power change is small, indicating that a maximum peak was not located, the cycles are detected by a tracking process with filtered data. This tracking process selects various lag values around the last two peaks identified in an attempt to select a lag value providing a large $r(\tau)$ peak which, if found, will provide a large power change. This reliably establishes the trigger point for the wide band autocorrelation calculations. The fine peak tracking is performed with wide band ECG data from the coarse (narrow band data) peak lag point in either direction until $r(\tau)$ maximums are determined within a resolution of 3 milliseconds. Thus, the final period determination is made with as much accuracy as the input sampling rate permits. If the subsequent cycle is not determined, its period will arbitrarily be set to the last period determined. Thereafter, if no cycles are determined, these data are considered to be noise. This procedure is performed economically since approximate peak locations are coarsely located, allowing the time-consuming determination of wide band, high resolution heart period to be made later within small time increments.

Filtering

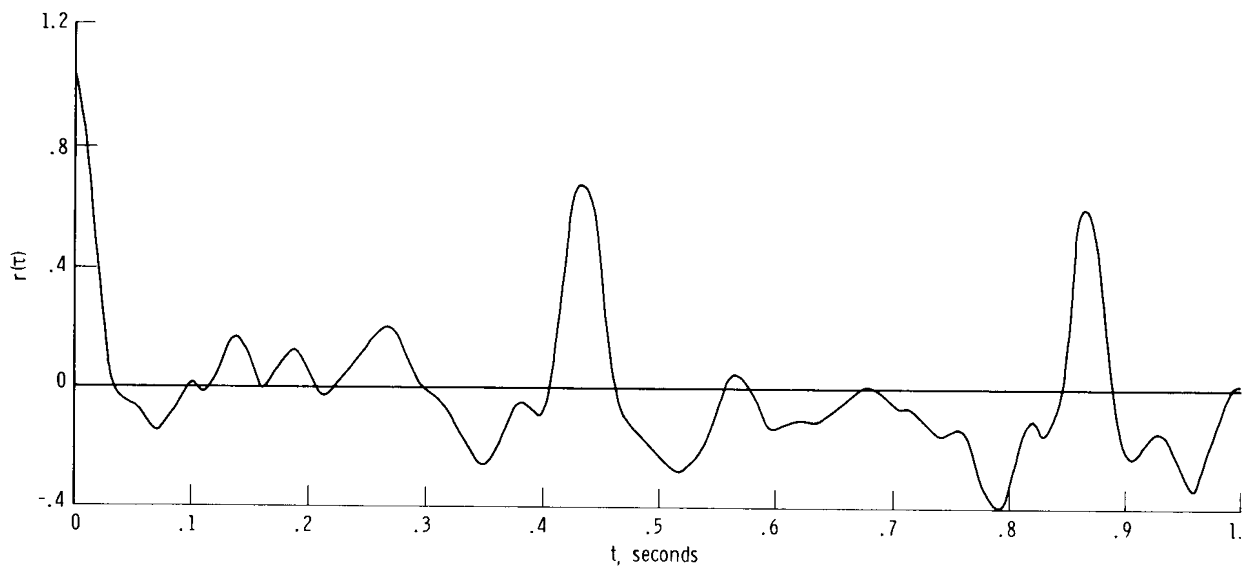
As mentioned, reduction of the bandwidth of the ECG prior to autocorrelation cardiotachometry was found to be necessary to create high reliability in the coarse peak (trigger point) detection mode. The use of digital filters accounts for approximately half the computer time required for each flight. At first it was not known whether digital filtering or analog filtering prior to digitizing would be more suitable during actual operations. Consequently, a study was conducted to evaluate the trade-offs relative to analog filtering and digital filtering.

The investigation of both filtering techniques indicated that the elimination of the higher harmonics resulted in waveform distortion as a function of phase shift. By using analog filtering, the flattening of the T-wave and the removal of the S-T segment became significant at high-pass cutoffs of 0.5 hertz and greater. However, one requirement for reliable autocorrelation cardiotachometry is that the slowly varying baseline shift be removed before autocorrelation. If the baseline were to shift periodically in the frequency range of interest at peak-to-peak values greater than those in the ECG waveform, erroneous maximum $r(\tau)$ values would be computed, which would invalidate the rate count. Data analysis indicated that baseline shift removal was not sufficient until high-pass cutoffs of 0.7 hertz or greater were achieved. This resulted in the obviously conflicting requirement of baseline shift removal and waveform preservation; however, digital filtering using a high-pass filter of 1 hertz with zero phase shift allowed the removal of the baseline shift and did not distort the waveform. To determine the optimum low-pass point, studies were conducted to establish the cleanest autocorrelation waveform and to provide the maximum amplitude difference between the maximum peak and subpeaks. The elimination of high frequencies was observed by noting the smoothing of autocorrelation waveforms with a lowering of high-frequency cutoff (figs. 3(a) to 3(c)). Eliminating the higher ECG harmonics widened the peaks centered at $\tau = 0$, 0.43, and 0.86 seconds as bandwidth was reduced, and blunted or rounded the autocorrelation peak tip as a result of the reduction of the low-pass point. Of particular interest was the increase in the amplitude difference between the subpeaks and maximum peak when a 20-hertz low-pass filter was used instead of a 40-hertz filter (figs. 3(a) and 3(b)). However, when a 5-hertz low-pass filter was used instead of a 10-hertz filter (figs. 3(c) and 3(d), the amplitude differences between the maximum peak and subpeaks decreased significantly. These trends were considered to be significant, because to establish reliable cardiotachometry the maximum peak amplitude must be significantly greater than any other peak amplitude. Further experiments with noise-degraded ECG data indicated that the largest difference between the maximum peak and subpeak amplitudes occurred at a low-pass frequency of 15 hertz.

The upper limit for the wide band ECG data was selected at 56 hertz to eliminate 60 hertz and higher frequency noise sources. For the wide band ECG data there was no change in reliability in autocorrelation cardiotachometry above 56 hertz with either

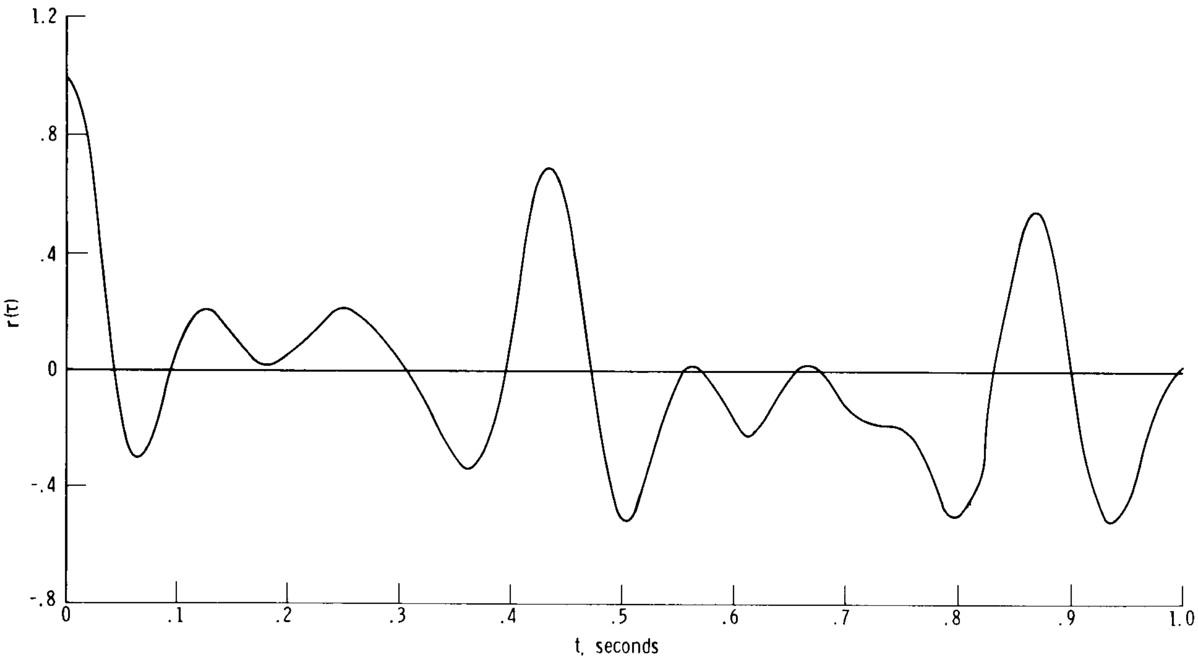


(a) Filter low pass = 40 hertz, heart rate = 138 beats per minute.

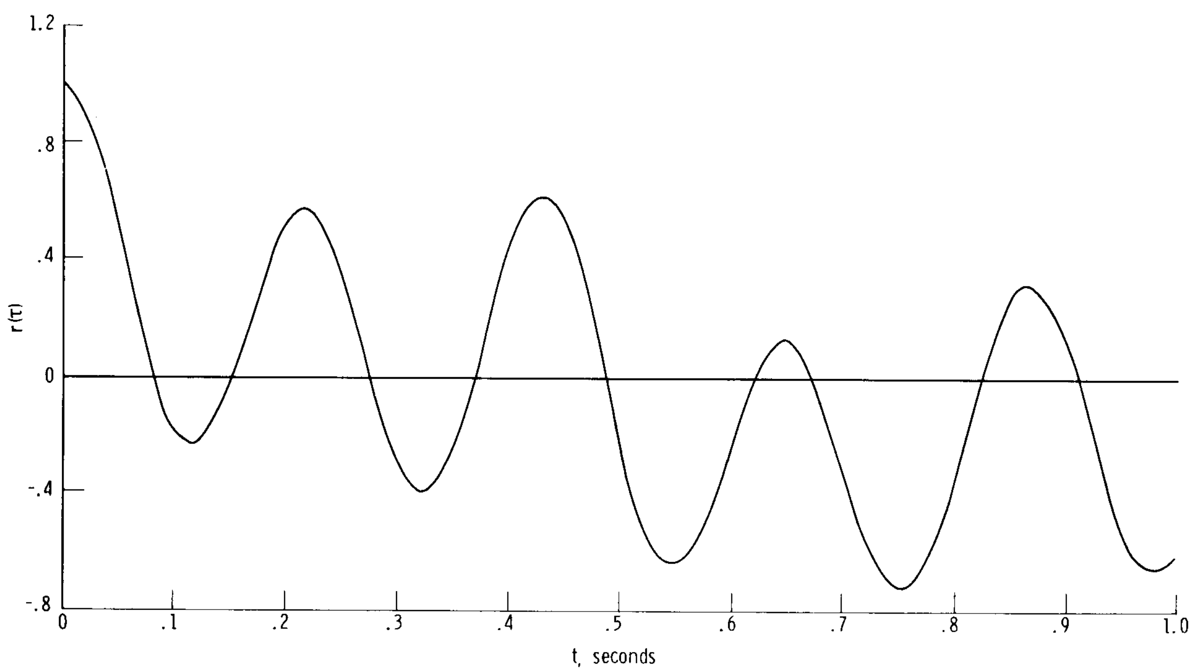


(b) Filter low pass = 20 hertz, heart rate = 138 beats per minute.

Figure 3. Normalized autocorrelation of ECG at various low-pass filters.



(c) Filter low pass = 10 hertz, heart rate = 138 beats per minute.



(d) Filter low pass = 5 hertz, heart rate = 138 beats per minute.

Figure 3. Concluded.

an analog or a digital filter. Consequently, analog filtering was chosen as the most economical means of filtering. The data were filtered between the multiplexer and analog-to-digital converters before digitizing (ref. 3).

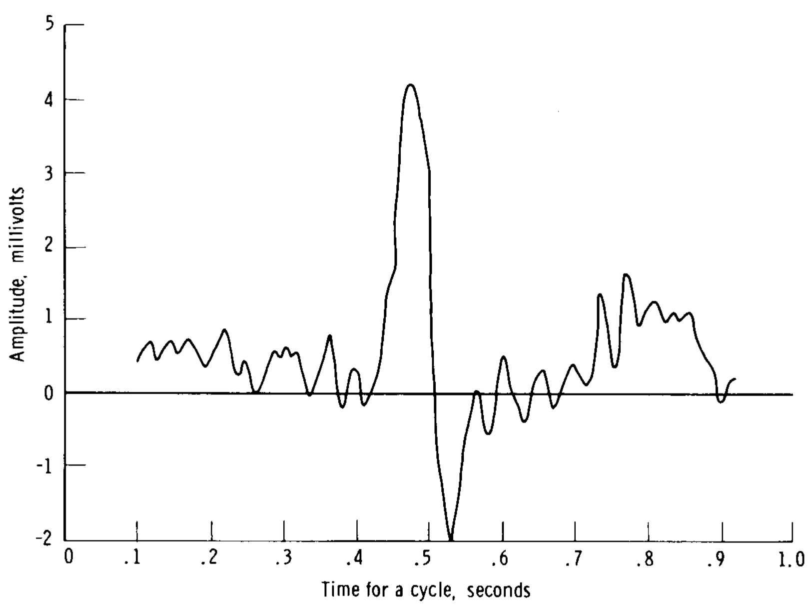
ECG WAVEFORM AVERAGING

Because it was desired to obtain time segment values within the ECG waveform, a technique was needed which would allow a stable, noise-free waveform. The decision was made to consider only typical cycles for each individual and to treat later the more complex problem of isolation and time segment analysis of aberrant beats. To achieve stable waveforms, averaging techniques were used to reduce the random noise power. The process studied was that of computing the arithmetic mean over N-waveforms of a noise-degraded signal.

If a signal with a noise power given by σ^2 is averaged, then N-waveforms will yield a noise power level equal to

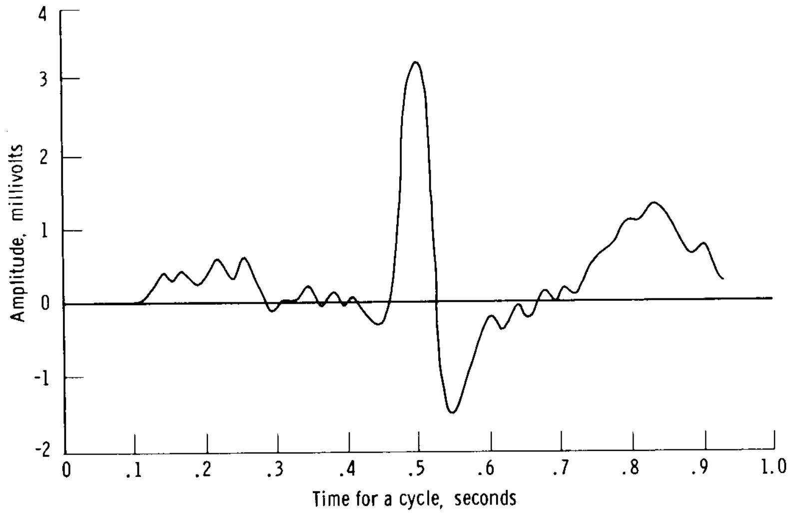
$$\sigma_{av}^2 = \frac{\sigma^2}{2N}$$

Thus, as shown in this equation, the greater the number of waveforms averaged, the less the noise power. Initially the ECG waveforms were phased by alining R-wave values through cross correlation of a "square" impulse with the current ECG waveform. Typical results of this study are shown in figures 4(a) and 4(b); figure 4(a) is a noisy ECG waveform, and figure 4(b) is a waveform from an average of 16 cycles. Although these results were encouraging, the waveform was not stable enough for time segment analysis.



(a) N = 1.

Figure 4. Typical averaged waveforms.



(b) N = 16.

Figure 4. Concluded.

At the conclusion of the autocorrelation cardiotachometry research, reliable waveform stability was achieved by averaging only those waveforms that autocorrelated at values of 0.8 or greater and by obtaining the ECG cycle period from the time difference between the maximum autocorrelation peak values. This part of the program removes the 0-millivolt baseline from the wide band data and performs the phasing on the S-wave portion of the QRS complex by detecting points on the highest frequency portion of the ECG cycle (S-wave) and alining these points. Stability was further enhanced by averaging cycles over a 1-minute duration, thus providing a significant reduction in noise power by considering a large number of cycles. These waveforms were considered to be stable (noise free) enough to apply time segment analysis techniques.

This final form of the averaging technique was incorporated into the main program through which all the data from the 1000-flight study were processed. These data are presented together with N, the number of cycles averaged, and T, the actual minute of the flight being considered, in figure 5. The value of N is not indicative of the heart rate because, as mentioned, only those values that autocorrelate above 0.8 are considered in the averaging process.

ECG PERIOD HISTOGRAMS

The heart rate versus the frequency of occurrence of each heart period is tabulated by the computer program and presented as 5-minute histograms throughout the flight, and a final histogram is plotted for the entire flight. A 5-minute histogram is shown in figure 6 with the normal average acceleration measured on the pilot (ref. 1). This program output also provides the number of suspected missed heartbeats during any 1-minute interval.

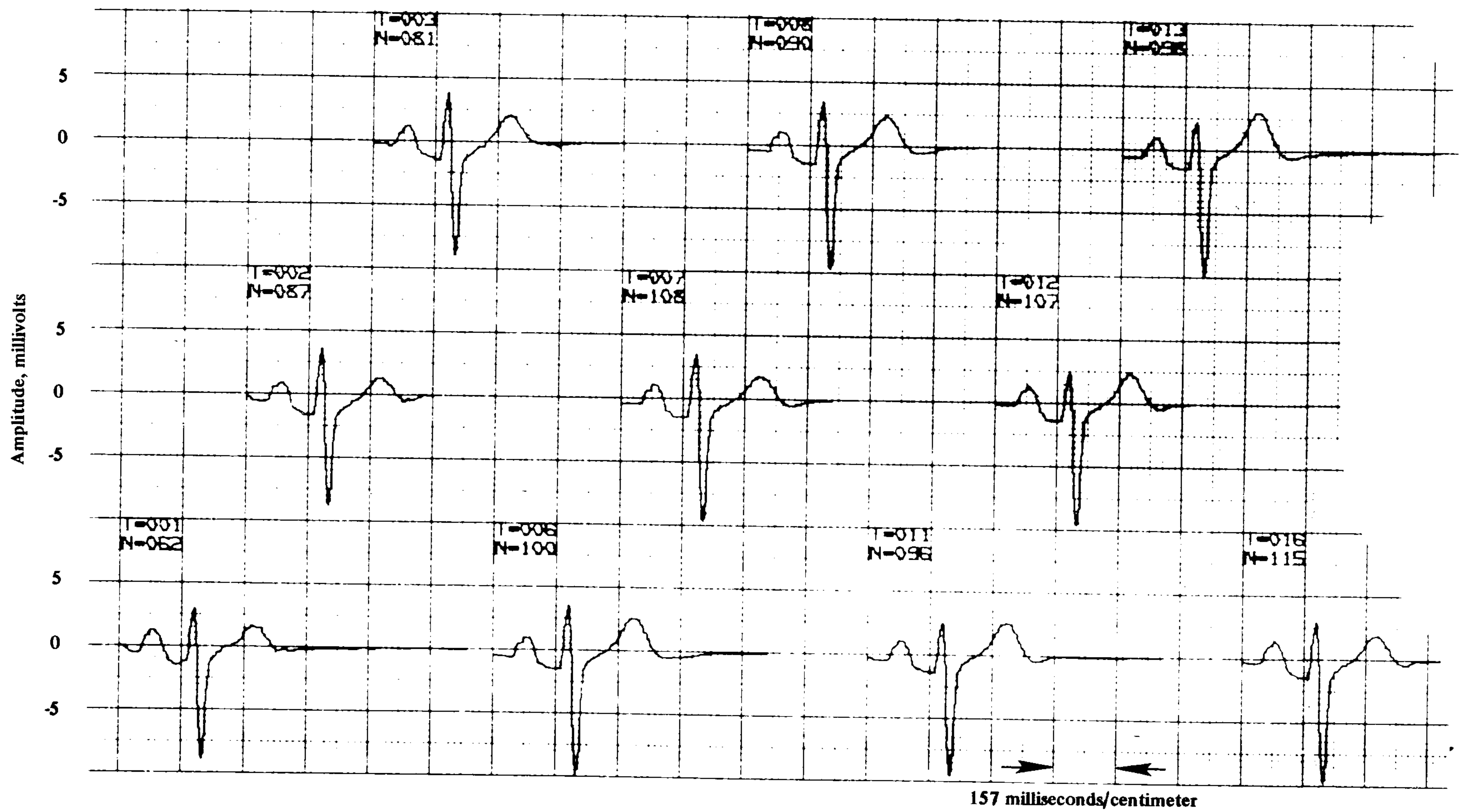


Figure 5. Digitally printed electrocardiographic averaged waveforms.

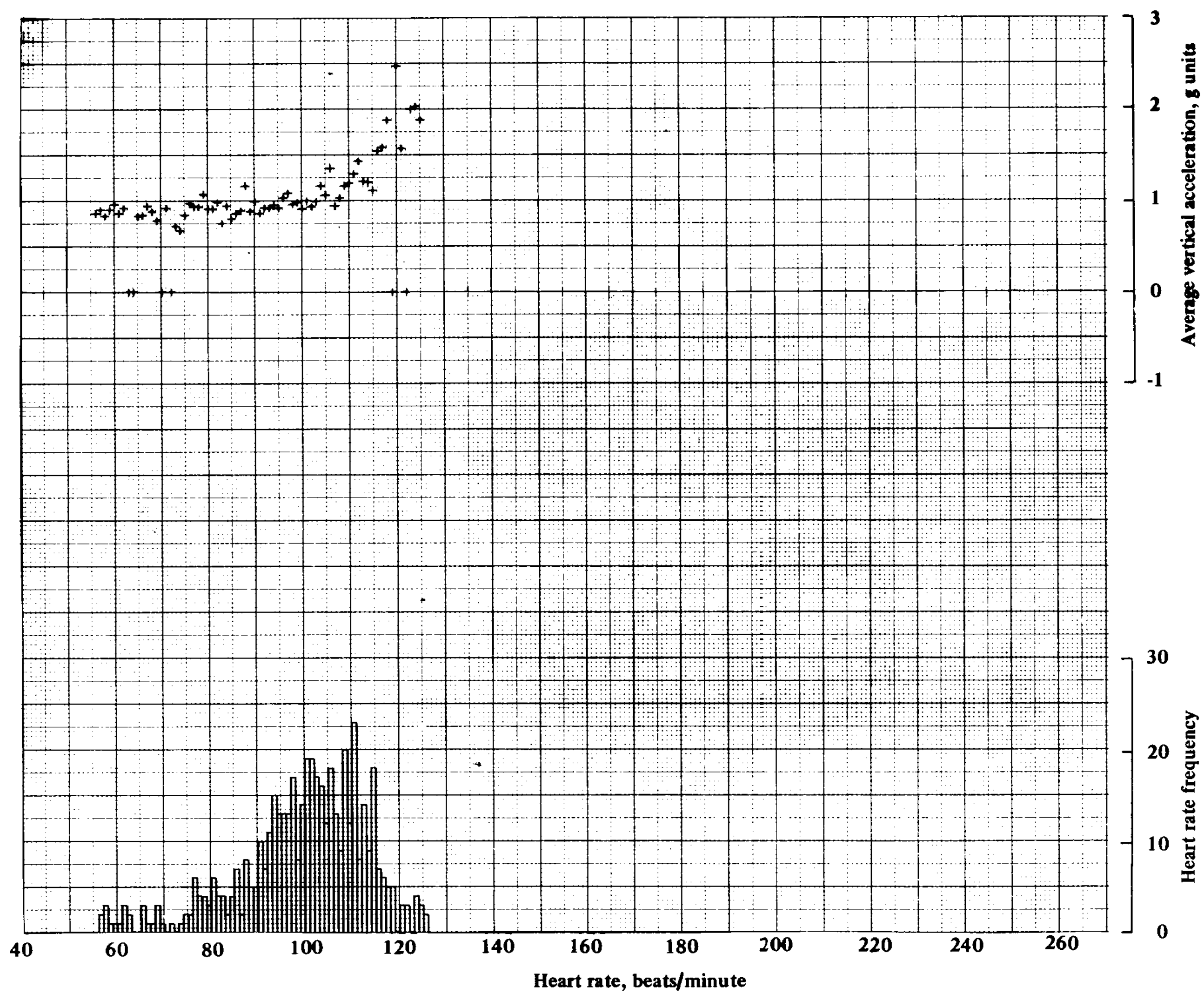


Figure 6. Digitally plotted 5-minute histogram of heart rate versus heart rate frequency plotted with average vertical acceleration measured on the pilot.

RESPIRATION RATE DETERMINATION

The technique of zero crossing detection was used to determine respiration rate monitored on pilots during flight (ref. 1). This technique, which entails counting the number of sign inversions in a given time period, is easily applied but requires a considerable amount of signal preconditioning to achieve reliability. To apply the technique it was necessary to use a low-pass filter with a corner frequency of 0.8 hertz. Extraneous signals within the low-pass range were for the most part excluded by including logic that allowed zero crossing counts only if the previous crossing was at least 720 milliseconds before, corresponding to a maximum of 43 breaths per minute. Small extraneous amplitude changes occurring after 720 milliseconds were further minimized by establishing the zero crossing threshold at 3 percent of the FM band edge, which established a lower digital amplitude boundary.

CONCLUDING REMARKS

Techniques were developed to automatically reduce a large amount of physiological data obtained during a 1000-flight biomedical study. Autocorrelation was found to be a successful method for determining heart rates in low signal-to-noise ratio environments. A two-step search routine (i.e., coarse, then fine search) was found to be an economical method of locating the maximum values of the autocorrelation function. Analog filtering was used for wide band filtering in the coarse search mode because it was more economical than digital filtering, but it distorted the waveform in the narrow band filtering. A digital filter which high-passed frequencies above 1 hertz and low-passed frequencies below 15 hertz gave good results in the narrow band, or fine search, routine. The selection of ECG waveforms that autocorrelated at values greater than 0.4 significantly enhanced the reliability and accuracy of heart period values.

The ECG waveform averaging technique provided well-defined and stable waveforms that may be used as a reference index for detecting aberrant beats.

The modified zero crossing technique was found to be a reliable and accurate technique for determining respiration rate in noisy environments.

Flight Research Center,
National Aeronautics and Space Administration,
Edwards, Calif., August 6, 1971.

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